A MULTI-LAYER FINITE ELEMENT MODEL OF THE SURFACE EMG SIGNAL

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Abstract- **The influence of skin, adipose tissue and bone on the rate of decay of the surface EMG signal around the limb was explored using a new finite element model. Replacing the outer layer of a homogeneous muscle model with a layer of highly resistive tissue, such as skin or fat, results in an increase in the surface potential. This also causes an increase in the rate of decay of EMG amplitude with increasing source depth and with increasing angular displacement from the source. EMG signals are examined as a bone is positioned at different locations throughout the muscle. Depending on its location, the highly resistive bone can significantly affect the amplitude of the surface potential. In a model of the upper arm, cross-talk around the limb was examined as subcutaneous tissue thickness was varied. EMG cross-talk was observed to increase with subcutaneous fat thickness. This is due to the relative increase in distance between source and recording site, rather than the material properties of the adipose tissue. The results illustrate the importance of including multiple tissue layers and imhomogeneities such as bone, when exploring aspects of surface EMG amplitude such as cross-talk.**

Keywords **- EMG, finite element model, cross-talk, subcutaneous tissue.**

I. INTRODUCTION

Many different models have been used to explore different aspects of the surface electromyographic (EMG) signal [1-5]. Such models have been predominantly based on analytical solutions to Laplace's equation. However, analytical models are limited to simple cylindrical or elliptical geometries and become increasingly difficult to handle as additional layers are incorporated. To simulate more physiological limb geometries, numerical methods such as finite element (FE) analysis must be employed [6]. A new FE model of the surface EMG signal is presented in this paper. The model incorporates skin, fat, muscle and bone tissue in an idealized cylindrical limb. The effect of material properties, inhomogeneous tissue and source location on the root mean square (RMS) amplitude and the median frequency of the EMG signal detected at different recording sites around the surface of the limb is examined. In particular, the relationship between subcutaneous tissue thickness and surface EMG cross-talk in the upper arm is explored.

II. METHODS

Five variations on the upper arm model are initially presented. In the first, the source is embedded in cylindrical, homogeneous muscle tissue of finite radial extent (model I). In the second and third models, the outer rim of the volume conductor is replaced with a layer of skin (model II), and then fat (model III), 2 mm and 5 mm thick respectively. A fourth model is composed of a 2mm thick skin layer, 3mm thick fat and muscle (model IV). A cylindrical core of cortical bone, radius 10mm, is then added to the model and the location of the bone is varied with respect to the active muscle fiber (model V). In models I-V, the total radius of the volume conductor is 50 mm. To examine changes in EMG cross-talk with variations in subcutaneous tissue thickness, action potentials are then generated with a multi-layer model composed of muscle, fat, skin and bone, as the thickness of the subcutaneous fat is varied, while keeping the volume of the muscle tissue constant. In this model, the radius from the center of the model to the surface of the muscle is 40 mm. The thickness of the subcutaneous fat layer is varied between 0 mm and 18 mm and a 1.3 mm skin thickness is assumed.

Using the FE method, very small, irregularly shaped elements are used to mesh areas of the volume conductor surrounding the region of interest, with a coarser resolution used in other regions of the model. The electric field variables are calculated at each node, located at the intersection of one or more elements. Unique material properties are defined for each element set. In accordance with most previous studies, the EMG models presented here are assumed to satisfy the conditions for quasi-stationarity [7]. This implies that the potential distribution in the conductor model may be determined from the unique solution of Laplace's equation that satisfies the specified boundary conditions. At the interface between any two media, the normal component of the current is assumed to be continuous. At the skin surface, it is assumed that the conductivity of the surrounding region (air), and hence the normal component to the field, is zero. The volume conductor is assumed to be of infinite length, grounded at infinity.

The transmembrane current density $I_m(z)$ is calculated from the analytic description of the transmembrane potential $V_m(z)$, presented in [8], assuming that $I_m(z)$ is directly proportional to the second derivative of $V_m(z)$ [1], [8]. $I_m(z)$, is applied to the surface of a 15mm long section of muscle fiber (diameter 100 μm) located within the conducting muscle tissue. The conductivity values for muscle σ_r, skin σ_s, infiltrated fat σ_f and cortical bone σ_b, used in the models presented here were obtained from parametric models presented in [9] at a frequency of 100Hz: σ_r =.2455 S/m; σ_s = 4.55x10^{-4} S/m; σ_f = 0.0379 S/m; σ_b = 0.02 S/m. Skin, fat and bone tissue are assumed to be isotropic. Muscle anisotropy (the ratio of muscle conductivity in the axial direction to muscle conductivity in the radial direction, σ_r) of 5 is assumed.

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Abstract

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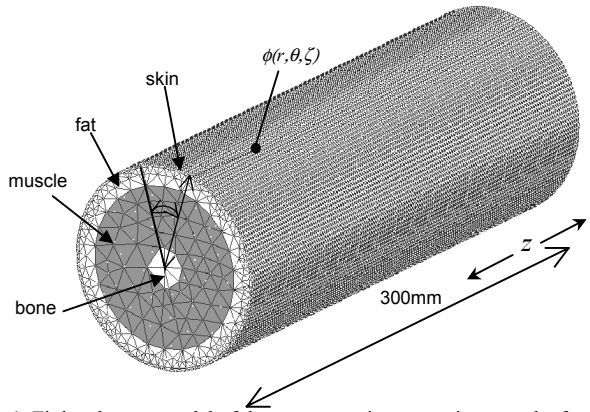

Fig. 1. Finite element model of the upper arm incorporating muscle, fat, skin and bone.

Flat, quadrilateral elements were first used to cover the skin surface with a uniform, high-resolution mesh. 3-dimensional tetrahedral elements were then used to mesh the interior regions of the model, Fig. 1. The size of the elements in the models ranged from 0.05mm, in the muscle fiber, to an element size of 12 mm. The finite element model was meshed and solved using EMAS software (Ansoft Corp., Pittsburgh, PA).

Simulated data is presented for bipolar recording electrodes, inter-electrode distance 20 mm, and assuming a muscle fiber conduction velocity (u) of 4 m/s. The spatially distributed waveform is interpreted as a function of time (t) using the relationship, $z = ut$.

III. Results

A. Effect of inhomogeneous layers

Action potentials detected along the surface of the volume conductor, directly above a 7 mm deep source, in models I-IV are compared in Fig. 2. It is apparent that replacing the outer layer of the model with a medium of low conductivity, such as skin or fat tissue, yields an increase in the amplitude of the surface EMG signal. An increase in the frequency content of the signal is also observed. The rate of decay of the surface potential, with increasing source depth and with increasing angular displacement from the source (θ), is also observed to increase. In Fig. 3 and Fig. 4, the RMS value and median frequency of the surface potentials directly above the source are presented as a function of increasing source depth.

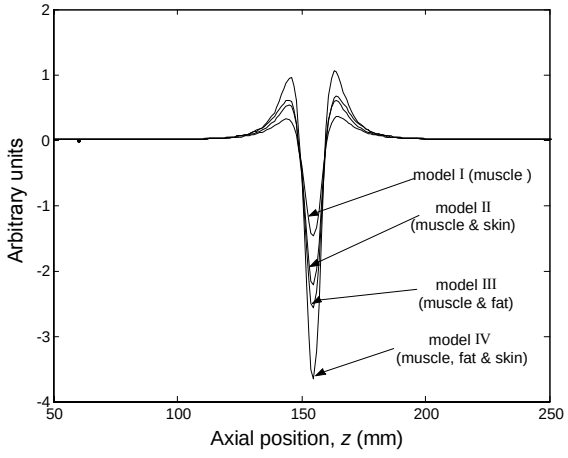

Fig. 2. Surface potential detected directly above a source of depth of 7 mm using models I-IV ($\theta = 0$).

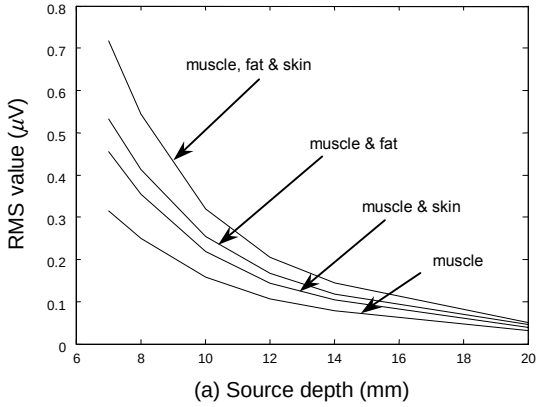

Fig. 3. Comparison of surface action potential RMS value with increasing source depth for increasingly complex models composed of muscle, skin and fat tissue.

B. Effect of bone

To examine the effect of bone on the amplitude of the EMG signal, the RMS value of the surface potential at different locations around the limb are calculated as the position of the bone with respect to a 10 mm deep muscle fiber is varied. The bone is initially located at the center of the model, then 2 mm below the active fiber. It is then moved to 2 mm below the muscle-fat interface and rotated 20°, 40° and 60° from the source. The five locations are illustrated in the inset, Fig. 5. Placing the bone at the center of the model (position 1) has little effect on the potentials detected at the skin surface, for the limb dimensions examined here. However, as the bone is moved closer to the source (position 2), an increase in the amplitude of the surface EMG signal is observed. When the bone is placed close to the muscle-fat interface (positions 3-5, Fig. 5) the potential at the surface increases in the region between source and bone, and also directly above the bone, Fig. 5. As the

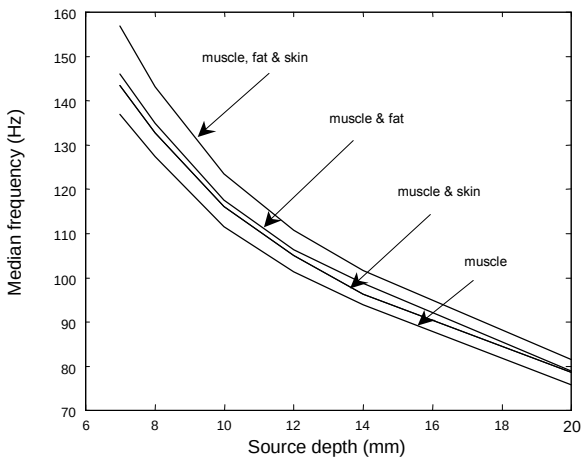

Fig. 4. Comparison of surface potential median frequency with increasing source depth for increasingly complex models composed of muscle, skin and fat tissue.

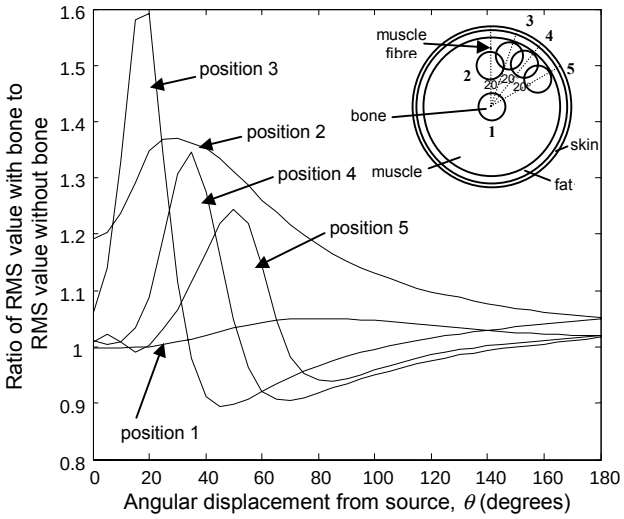

Fig. 5. Effect of bone on surface potential RMS amplitude for different bone locations (1-5 indicated in inset). RMS values with bone are normalized with respect to the values calculated without bone at each angular displacement from the 10 mm deep source, θ.

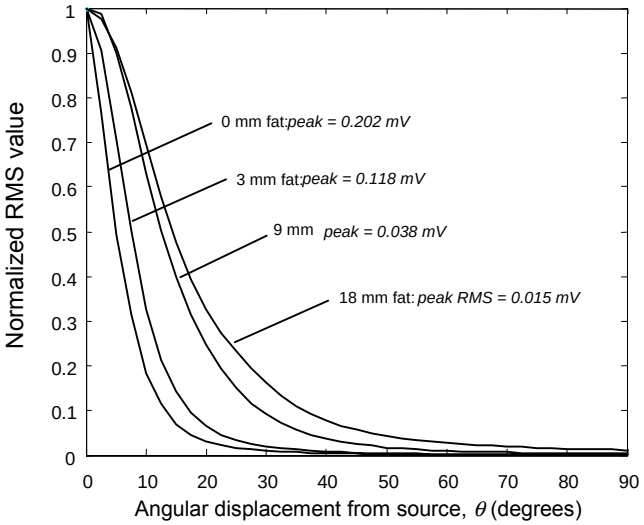

Fig. 6. RMS value of the surface action potential detected around the volume conductor, normalized with respect to the RMS value detected located 3 mm below the surface of the muscle tissue, with subcutaneous fat tissue of thickness 0, 3, 9 and 18 mm. directly above the source (in italics).

detection point is moved away from the bone, surface potentials lower than those detected in the skin, fat and muscle model are observed.

The RMS values of the action potentials calculated around the surface of the volume conductor, for each bone location, are compared in Fig. 5. At each angular position, θ, the RMS value is normalized with respect to the RMS value from model IV, where bone is not included.

C. Effect of subcutaneous tissue thickness

To examine the effect of variations in subcutaneous fat thickness on surface EMG cross-talk under more physiological conditions, the EMG signal detected at the skin surface is examined as the thickness of the subcutaneous fat layer is varied, keeping the position of the source and the volume of the muscle tissue constant. Cross-talk at a given angular displacement from the source is estimated as the RMS value of EMG signal normalized with respect to the value directly above the source. Results are presented for bipolar recording electrodes, inter-electrode distance 20 mm. At all source depths, increasing the thickness of the subcutaneous fat results in a decrease in the amplitude of the surface EMG signal at all points around the limb and a simultaneous increase in EMG cross-talk. In Fig. 6, results are presented for a source located 3 mm below the surface of the muscle-fat tissue interface. The rate of decay of the action potential RMS amplitude around the limb is presented for each subcutaneous fat thickness. Peak RMS values (at $\theta = 0$) are given in italics.

IV. Discussion

Replacing the outer layer of the homogenous muscle model with a highly resistive material such as fat and/or skin, was found to cause an increase in the amplitude and frequency content of the potential observed at the surface, Figs. 3 and 4. This perhaps counter-intuitive result is explained by the decrease in current entering the outer annulus as its resistance increases, causing the potential dropped across the highly resistive outer layer to decrease. The potential observed at the surface is then closer to the potential distribution at the muscle-fat interface, yielding the increase in amplitude and frequency content observed at the skin surface. As a result of this 'focusing effect' of the high resistance layers, the rate at which the RMS amplitude at the surface decays with increasing source depth increases. These observations are in agreement with the observations made by Roeleveld *et al.*, [10], in an analytical multi-layer EMG model. The value of skin conductivity used in [10] however, is substantially higher than the values reported by Gabriel *et al.*, [9], being higher that the conductivity of either muscle or fat, and tends to reverse the effect of the highly resistive subcutaneous tissue.

Depending on the location with respect to the source and the recording electrodes, highly resistive regions such as bone, can have a significant effect on the potential observed at the surface. When compared with the homogeneous skin, fat and muscle model, an increase in the surface potential is observed above regions where current is constrained by the inhomogeneity, between source and bone, and directly above the bone. In the region to the other side of the bone, where current flow has been blocked, a decrease in potential is observed, Fig. 5. The magnitude of this effect depends on several factors, including the location of the inhomogeneity with respect to the source and the boundary of the volume conductor and the location of the recording electrodes. In the upper arm, the humerus is approximately located at the center of the limb and has little effect on simulated surface potentials. However, at other locations, where the bone is located close to the muscle of interest, the effect on the surface EMG signal may not be negligible and should be considered.

Several experimental studies have noted an increase in surface EMG cross-talk with increasing skin-fold thickness

[11], [12]. The reasons behind these observations are not immediately obvious [11]. It is not clear whether the increase in EMG cross-talk observed is simply due to greater spatial filtering of the signal as the distance between source and electrode increases, or whether the material properties of the subcutaneous tissue cause an additional attenuation of the signal. In this study, surface EMG cross-talk was examined for several source depths as subcutaneous fat thickness was increased. The simulation results demonstrate an increase in cross-talk with subcutaneous tissue thickness, accompanied by decrease in EMG amplitude at all sites, in agreement with these experimental observations. From the simulation studies it is concluded that the experimentally observed increase in surface EMG cross-talk with increasing subcutaneous fat thickness is due to the increased distance between recording site and muscle fibers, rather than any filtering properties of the adipose tissue. Indeed, were the conductivity of the fat tissue equal to that of muscle, even greater levels of surface EMG cross-talk would be expected.

For simplicity, only single fiber action potentials have been considered. The influence of motor unit firing rates, synchronization, motor unit and fiber distribution must also be considered when interpreting voluntary surface EMG signals. Nevertheless, the single fiber model is a convenient method of exploring issues such as volume conduction through inhomogeneous media and variations in subcutaneous fat thickness without the further complicating effects related to motor control. All of the results presented here are for a standard single differential electrode configuration with an inter-electrode distance of 20 mm. The recording electrode configuration is known to play an important role in determining levels of surface EMG cross-talk. The sensitivity of the surface electrode pick-up range to electrode configuration has been discussed in detail elsewhere and is therefore not addressed here. It has also been assumed that at the frequencies of interest the biological tissues included in the model behave as if they are purely resistive. While the values of muscle conductivity and permittivity presented in [9] satisfy the requirements of a purely resistive homogeneous model as described by [7], the influence of skin and fat capacitance in the inhomogeneous model is more difficult to predict and requires further exploration.

Numerical methods, including finite element modeling, open up the possibility of developing complex EMG models based on realistic geometries, which are beyond the scope of analytical methods widely employed. Although the models presented here are based on an idealized cylindrical limb, it is anticipated that the next generation of models will incorporate real anatomical data derived using magnetic resonance imaging techniques. The results presented illustrate the importance of addressing skin and subcutaneous fat thickness, the location of bone and other inhomogeneities when considering aspects of EMG amplitude such as cross-talk.

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